COMPUTER SIMULATION OF IMPACT TEST AND INJURY CRITERIA ANALYSIS BY USING SENSITIVITY ANALYSIS

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ABSTRACT

Articulated multibody mechanical model of human body is used as a tool to investigate injury mechanism during a car crash event. Dynamic response of the proposed multibody system is calculated for two sets of human body mechanical parameters. First set of mechanical parameters is generated by GEBOD (GEnerator of BOdy Data) from Wright Patterson Air Force Base (WPAFB) and in this research represents standard dummy. The second set of mechanical parameters was obtained from measurements of human body parts conducted on real human corpses. On the basis of both dynamic responses some injury criteria are calculated and discussed.

INTRODUCTION

The design of vehicle and vehicle safety equipment have been for many years evaluated with respect to safety for dummies. Dummies represent mechanical substitute of real humans. The design and mechanical properties of dummies such as geometry, inertia properties etc. do not conform to individual real human values. In fact they are calculated as certain statistic average of male or female human population. For frontal impact tests two Hybrid III dummies are used last few years (NHTSA, 1997), known as 95th percentile (represents large male) and 5th percentile (represents small female). For side impact tests SID, EUROSID-1 and BIOSID dummies have been used. Also a small female dummy for side impact has been introduced in 1995 (Daniel et al., 1995). A large number of impact simulations with such different dummies have shown a wide range of possible dynamic response for the same type of external load. It is obvious that due to different mechanical properties and also different initial position of human body, the dynamic response and injuries exceed the range of expected results. Therefore it is necessary to study how human body variability in geometry, proportion, inertia properties and initial position influence the dynamic response and possible injuries. Due to the time and costs, only a limited number of new dummies that could provide useful answers will be used in the future. Also one could conclude that is impossible to construct a dummy that would represent all variations in geometry and corpulence of human population. It seems that the only possible approach, that is capable to address this problem, is computer simulation of impact test on the basis of suitable mechanical and mathematical models of human body.

Till now limited attention is directed to question concerning the effect of human body variable anthropometry on the dynamic response. One of the known approaches to address this questions is scaling model (MADYMO, 1998). The idea of this approach is to generate a set of target mechanical parameters from a relevant human population and then scale existing values towards desired individual anthropometry. Following the work of (Happee, 1998) also some other mechanical parameters are scaled. In fact there is scaling of:

- geometry,
- mass and inertia tensor,
- all joint characteristic (stiffness, friction, damping, hysteresis),
- ellipsoids and contacts characteristics,
- all other force models,
- sensor location,
- reference length for the V*R criterion.

Usually empirical regression equations, based on extensive anthropometric measurements on relevant human population, are used for predicting a set of unknown human body mechanical parameters from known ones. One of common known sets of regression equations is used by GEBOD (Wright Patterson Air Force Base, 1994). There are four groups of regression equations that are used to determine the human body joint location coordinates, segment volumes, segment masses and inertia tensors. Each regression equation is a first order linear algebraic equation involving either standing height or body weight, or both of them as the independent variables.

MECHANICAL MODEL OF HUMAN BODY

The anthropomorphic multibody system used to represent the human body in this research is depicted on Fig. 1. The multibody system conforms to standard ATB (Articulated Total Body model) (Cheng, 1998) body setup and could therefore use mechanical parameter values generated by GEBOD. Multibody system consists of 17 elements connected with revolute and spherical kinematic pairs representing human joints (Kapandji, 1990).

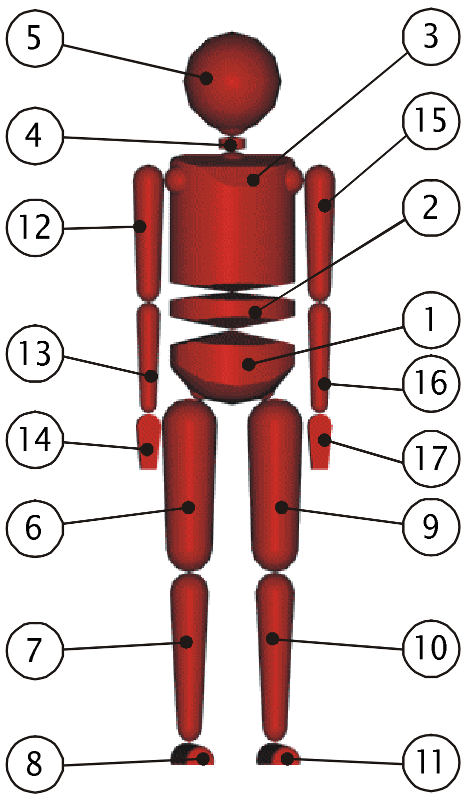

Figure 1.

Anthropomorphic multibody system.

The head-neck joint is represented with spherical kinematic pair (3DOF), the neck-thorax joint with revolute kinematic pair (1DOF), the shoulders with spherical kinematic pair (3DOF), the elbow with revolute kinematic pair (1DOF), the wrist with spherical kinematic pair (3DOF), the lower trunk joint with spherical kinematic pair (3DOF), the hip with spherical kinematic pair (3DOF), knees with revolute kinematic pair (2DOF) and ankles with revolute kinematic pair (2DOF). General multibody simulation programs (Kwatny, 1995) are capable of generating non-linear differential equations of motion for such a multibody system in the well-known Poincare's form

$$\dot{\vec{q}} = \mathbf{V}(\vec{q})\vec{w} \tag{1}$$

$$\begin{aligned} &\mathbf{M}_w(\vec{q})\dot{\vec{w}} + \mathbf{C}_w(\vec{q},\vec{w})\vec{w} + \mathbf{F}_w(\vec{q}) = \mathbf{Q}_w \\ &\vec{q}(0) = \vec{q}_0, \quad \vec{w}(0) = \vec{w}_0, \\ &t \in [0,\tau] \quad . \end{aligned} \tag{2}$$

The equations of motion (1) and (2) are Lagrange's equations

$$\begin{aligned} &\mathbf{M}\ddot{\vec{q}} + \mathbf{C}\dot{\vec{q}} + \mathbf{F}\vec{q} = \mathbf{Q} \\ &\vec{q}(0) = \vec{q}_0, \quad \dot{\vec{q}}(0) = \dot{\vec{q}}_0, \\ &t \in [0,\tau], \end{aligned} \tag{3}$$

if $\mathbf{V}(\vec{q}) = \mathbf{I}$, where $\mathbf{I}$ is identity matrix. In this case quasi velocity vector $\vec{w}$ is equal to time derivatives of generalized coordinates $\dot{\vec{q}} = \vec{w}$.

MECHANICAL PROPERTIES OF HUMAN BODY

One set of geometrical and inertial properties is obtained from GEBOD. Body height and body weight, as measured on the test subject, are used as input into GEBOD.
Another set of geometrical and inertial properties of body parts is obtained from measurements on human corpses in accordance with ethical and legal provisions. Corpses were fixed in 1% phenol-formaldehyde solution, which provided rigidity of body parts as well as of entire body. The first measurement is done on the body part consisting of head, neck and trunk as one unit. Next, this unit is divided into four or five sub-units. The torso consists of thorax (3), abdomen (2) and pelvis (1). Other subunits are head (5) and neck (4). A measurement is done on each of these separate sub-units. Head with neck is dissected from trunk at the level of the C7-Th1 intervertebral disc. The dividing plane is inclined forwards and goes through jugular notch and sternoclavicular joints at the front of the trunk. Furthermore, head and neck are separated through the oblique plane joining atlantooccipital joint and hyoid bone. Thorax and abdomen are divided dorsally through the plane lying in the intervertebral disc Th10 - Th11. The plane follows obliquely the lower border of the 10th rib to reach its lowest point; from there it runs in the subcostal plane. Abdomen and pelvis are divided through the intervertebral disc L3-L4, following the iliac crest obliquely towards the anterior abdominal wall. Extremities are dissected from trunk in shoulder and hip joints. Muscles that

embrace joints as well as ligaments and synovial membranes are cut. Muscles of the shoulder ring and shoulder blades are therefore considered part of thorax; muscles of pelvic ring are mostly considered part of pelvis. Extremities are first measured as whole units each consisting of three parts, then they are dissected through joint crevices to get three separate parts from each of them. The upper extremity consists of upper arm (12, 15), lower arm (13, 16) and hand (14, 17), the lower extremity consists of upper leg (6, 9), lower leg (7, 10) and foot (8, 11). Upper extremity is dissected through elbow joint and wrist joint, lower extremity is dissected through knee joint and the upper ankle joint.

Components of inertia tensor J and the respective neutral axis offsets e, defining centre of gravity needed by the mathematical model are calculated for each body part from its mass m, the distance between its axes of oscillation d_{AB} and its oscillation periods t_A and t_B, by means of system of equations

$$e = \frac{4{d_{AB}}^2\pi^2 - g \cdot d_{AB} \cdot t_B^2}{8d_{AB}\pi^2 - g\left(t_A^2 + t_B^2\right)} \tag{4}$$

$$J = -e^2 m + \frac{e \cdot g \cdot m \cdot t_A^2}{4\pi^2} \tag{5}$$

A measuring device depicted on Fig. 2 has been developed.

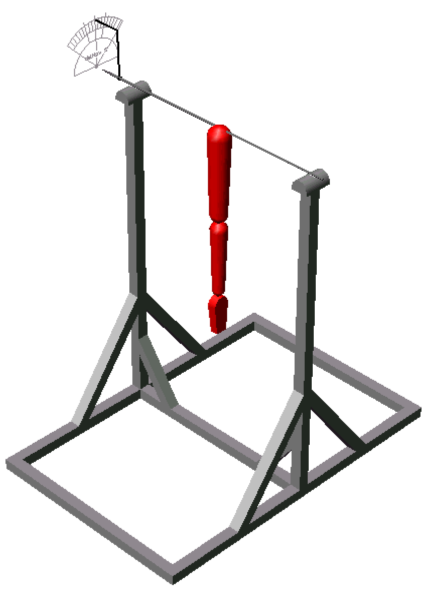

Figure 2.

Measuring device.

Device consisting of a horizontal frame onto which two vertical beams are attached. At the top of each vertical beam there is a seat made of cylindrical steel rod. The body part to be measured is then attached onto a square-section stainless steel beam so that its axis of rotation is perpendicular to the plane of oscillation. The stainless steel beam with attached body part is put onto the cylindrical seat so that it lies freely on one of its edges. A notch in the seat prevents the beam with the body part from sliding off its position during oscillation. For each body part, oscillation times in two support points (proximal and distal) and two planes (frontal and sagital) are measured. After the stainless steel beam is removed from the body part, the part is weighed; its absolute length is measured as well as the distance between support points in both planes. Three measurements are done for each plane and each support point, resulting in 12 measurements per body part. To achieve oscillation, each body part is declined 20° off its gravitational position and then released to freely oscillate. Amplitude is measured with goniometer attached to the stand. The time is measured, in which 30 or 50 full oscillations (depending on mass and length of body part) are completed. For each body segment calculated physical quantities are evaluated and compared to eliminate possible errors.

The two sets (from measurement and from GEBOD) of test subject mechanical properties are summarized in Table 1, appended at the end of paper.

INJURY CRITERIA AND SENSITIVITY ANALYSIS

Presently, there are several physical parameters used in the evaluation of human injury. The most important are the impact force, the impact duration, the kinetic energy, the displacement and the acceleration level. The critical values of considered parameters are obtained experimentally with animal subjects, cadavers, volunteer subjects and dummies. Several injury predictive approaches (Mercedes Benz AG, 1992) have been developed on the basis of experimental results. Two of the well known are Severity Index (SI) and Head Injury Criteria (HIC), which are postulated as functional

$$SI = \int_{t_1}^{t_2} a^{2.5}(t)dt\ , \tag{6}$$

$$HIC = \left\{ (t_2 - t_1)\left[\frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} \right\}_{MAX.} . \tag{7}$$

The general form of (6) and (7) could be represent as

$$\Psi = \int_0^\tau h(\vec{b},\vec{q},t)dt\ , \qquad (8)$$

where $\vec{b}$ is vector of the mechanical parameters and $\vec{q}$ is vector of generalized coordinates, describing the motion of the segments of anthropomorphic multibody system. The effect of human body variable anthropometry on the dynamic response and criteria's such as (8) could be mathematically described by calculated variation (total derivative) of the functional (8) subject to vector of mechanical parameters $\vec{b}$.

Sensitivity analysis is a procedure for calculating response variations of a functional that depends implicitly on the system parameter vector. Implicit dependence arises due to a closed-form solution of the physical laws that govern the system response for a system parameter vector. Variation of functional (8) for a finite dimensional system can be calculated (Arora, 1992)

$$\bar{\delta}\Psi \equiv \frac{d\Psi^T}{d\vec{b}}\delta\vec{b} = \left(\frac{d\Psi^T}{d\vec{b}} + \frac{d\Psi^T}{d\vec{q}}\cdot\frac{d\vec{q}}{d\vec{b}}\right)\delta\vec{b}, \qquad (9)$$

where $\bar{\delta}$ represents the total design variation operator. For complex mechanical systems it is inefficient to compute matrix $\frac{d\vec{q}}{d\vec{b}}$ appearing in (9) directly. This computation could be avoided with the help of adjoint variable method.

Let $\vec{q}_a$ represent kinematically admissible adjoint vector such, that the augmented functional

$$L = \int_0^\tau h(\vec{b},\vec{q},t)dt + \int_0^\tau \vec{q}_a^{\,T}\left(\mathbf{M}\ddot{\vec{q}} + \mathbf{C}\dot{\vec{q}} + \mathbf{F}\vec{q} - \mathbf{Q}\right)dt \qquad (10)$$

is stationary with respect to variations of the $\vec{q}$. Defining

$$\bar{h} := h + \vec{q}_a^{\,T}\left(\mathbf{M}\ddot{\vec{q}} + \mathbf{C}\dot{\vec{q}} + \mathbf{F}\vec{q} - \mathbf{Q}\right), \qquad (11)$$

expression (10) could be rewrite as

$$L = \int_0^\tau \bar{h}dt\ . \qquad (12)$$

If we define the explicit parameter variation operator $\tilde{\delta}$ the equality of the total parameter vector of the functional (8) is given as the explicit parameter vector variation of augmented functional (10)

$$\bar{\delta}\psi = \tilde{\delta}L\ . \qquad (13)$$

Equality (13) enable one to calculate $\vec{q}_a$ by solving nonlinear diferential equations

$$\begin{aligned} &\mathbf{M}\ddot{\vec{q}}_a + \mathbf{C}\dot{\vec{q}}_a + \mathbf{F}\vec{q}_a = -h,_q \\ &\vec{q}_a(\tau) = 0, \quad \dot{\vec{q}}_a(\tau) = 0, \end{aligned} \qquad (14)$$

and finally calculate the total variation (total derivative) of the functional (8) as

$$\bar{\delta}\psi = \int_0^\tau \left\{h,_b + \left[\left(\mathbf{M}\ddot{\vec{q}} + \mathbf{C}\dot{\vec{q}} + \mathbf{F}\vec{q} - \mathbf{Q}\right),_b\right]^T \vec{q}_a\right\}\delta b \cdot dt\ . \qquad (15)$$

EXAMPLE

A numerical example demonstrates the difference in head acceleration over time between the two sets of data for the same test subject (adult male, F_G=539 N, h_{stand}=1.7m), the first one obtained from measurement and the second one generated by GEBOD. The two curves (Fig. 3) are calculated with the same mathematical model and same external load (Cheng et. al. 1998)

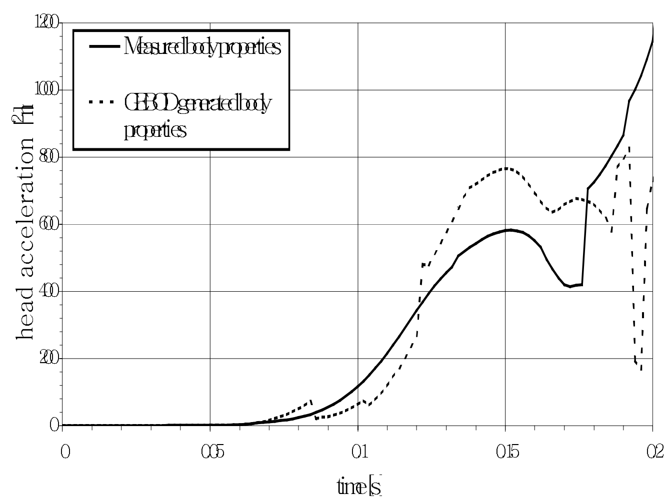

Figure 3.

Head acceleration.

A reduced form of (8) is used

$$\Psi = a(t)\,, \tag{16}$$

where $a(t)$ represent head acceleration. Time history of the head acceleration sensitivity coefficient with respect to the mass of the head (Fig. 4) has been calculated. Sensitivity coefficient shows significant influence of the head mass variation on its acceleration.

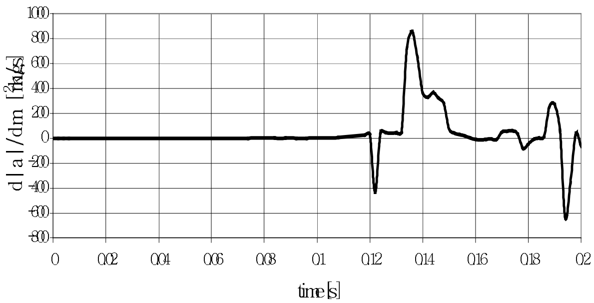

Figure 4.

Head acceleration sensitivity.

CONCLUSIONS

Research confirms that the same crash scenario has different effect on different size test subjects. The sensitivity analysis is proven as a suitable tool to perform quantitative analysis of the effect of human body geometrical and inertial properties on the dynamic response. Sensitivity analysis could provide useful information to procedures such as occupant model scaling. Also, it provides necessary information to optimize vehicle safety with the help of optimization methods based on non-linear programming problem formulation (Hsieh 1984).
The adjoint variable approach of sensitivity analysis is strongly recommended for complicated structures such as human body and vehicle restraint systems.

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APPENDIX

Table 1.
Comparison of human body segment data obtained from GEBOD and from measurement

SEGMENT		GEBOD			MEASUREMENT		
		WEIGHT [N]	MOMENT OF INERTIA [kgm^2]		WEIGHT [N]	MOMENT OF INERTIA [kgm^2]	
i	SYMBOL		J_x	J_y		J_x	J_y
1	LT	65,68	0,0371	0,0430	78,480	0,0739	0,0499
2	CT	14,53	0,0063	0,0027	58,860	0,254	0,235
3	UT	154,04	0,2377	0,1682	147,150	0,2435	0,1899
4	N	7,22	0,0009	0,0011	10,045	0,0023	0,0042
5	H	39,36	0,0193	0,0220	34,335	0,0122	0,0161
6	RUL	67,93	0,0982	0,1024	48,658	0,0740	0,0715
7	RLL	28,44	0,0401	0,0406	19,620	0,0243	0,0243
8	RF	7,42	0,0033	0,0031	7,652	0,0027	0,0025
9	LUL	67,93	0,0982	0,1024	48,658	0,0740	0,0715
10	LLL	28,44	0,0401	0,0406	19,620	0,0243	0,0243
11	LF	7,42	0,0033	0,0031	7,652	0,0027	0,0025
12	RUA	12,03	0,0063	0,0063	16,677	0,0118	0,0123
13	RLA	9,25	0,0053	0,0054	8,204	0,0039	0,0039
14	RH	3,99	0,0010	0,0008	3,924	0,0008	0,0005
15	LUA	12,03	0,0063	0,0063	16,677	0,0118	0,0123
16	LLA	9,25	0,0053	0,0054	8,204	0,0039	0,0039
17	LH	3,99	0,0010	0,0008	3,924	0,0008	0,0009